Reduced ascending aorta distensibility relates to adverse ventricular mechanics in patients with hypoplastic left heart syndrome: Noninvasive study using wave intensity analysis

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Objective: To evaluate the aortic arch elastic properties and ventriculoarterial coupling efficiency in patients with single ventricle physiology, with and without a surgically reconstructed arch.

Methods: We studied 21 children with single ventricle physiology after bidirectional superior cavopulmonary surgery: 10 with hypoplastic left heart syndrome, who underwent surgical arch reconstruction, and 11 with other types of single ventricle physiology but without arch reconstruction. All children underwent pre-Fontan magnetic resonance imaging. No patient exhibited aortic recoarctation. Data on aortic wave speed, aortic distensibility and wave intensity profiles were all extracted from the magnetic resonance imaging studies using an in-house–written plug-in for the Digital Imaging and Communications in Medicine viewer OsiriX.

Results: Children with hypoplastic left heart syndrome had significantly greater wave speed ($P = .002$), and both stiffer ($P = .004$) and larger ($P < .0001$) ascending aortas than the patients with a nonreconstructed arch. Aortic distensibility was not influenced by ventricular stroke volume but depended on a combination of increased aortic diameter and abnormal wall mechanical properties. Those with hypoplastic left heart syndrome had a lower peak wave intensity and reduced energy carried by the forward compression and the forward expansion waves, even after correction for stroke volume, suggesting an abnormal systolic and diastolic function. Lower wave energy was associated with an increased aortic diameter.

Conclusions: Using a novel, noninvasive technique based on image analysis, we have demonstrated that aortic arch reconstruction in children with hypoplastic left heart syndrome is associated with reduced aortic distensibility and unfavorable ventricular-vascular coupling compared with those with single ventricle physiology without aortic arch reconstruction. (J Thorac Cardiovasc Surg 2012;144:1307-14)

The first-stage palliative procedure for the treatment of hypoplastic left heart syndrome (HLHS) or Norwood procedure[1] requires extensive surgical reconstruction of the aortic arch with augmentation, typically using a pulmonary homograft patch.[2] Despite acceptable short-term outcomes,[3] long-term concern exists regarding the function of the right ventricle in a systemic position. Recent evidence suggests that Fontan patients with a systemic, single, right ventricle are at increased risk of ventricular dysfunction[4] and reduced exercise capacity[5] in the mid to long term. Although progressive attrition is thought to be related to abnormal ventricular structure, or even inherent to the Fontan physiology, concern exists that increased afterload could be the cause of premature ventricular dysfunction.[6] For example, in this setting, aortic coarctation is known to lead to ventricular dysfunction.[7] However, it is now becoming evident that ventricular afterload can be increased even by mechanisms different from vascular stenosis, such as increased aortic input and characteristic impedance and increased wave reflection.[8,9]

Additional surgical factors could play an important role in determining the increased ventricular afterload. Recent evidence has suggested that patients with HLHS have reduced distensibility of the ascending aorta[10-12] as a consequence of extensive reconstruction of the aortic arch at the Norwood operation. However, the relationship between the abnormal elastic properties of the proximal aorta and aortic arch and ventricular mechanics in this setting has not been assessed in full.

The aim of the present study was to noninvasively assess the mechanical properties of the ascending aorta in patients with single ventricle physiology after superior bidirectional cavopulmonary anastomosis and to evaluate the influence of

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Abbreviations and Acronyms

FCW = forward compression wave
FEW = forward expansion wave
HLHS = hypoplastic left heart syndrome
LV = left ventricular
MRI = magnetic resonance imaging
SV = stroke volume

abnormal aortic mechanical properties on ventricular function and the efficiency of vascular-ventricular coupling.

METHODS

Patients

We selected 21 children (mean age, 4.0 ± 1.4 years) with single ventricle physiology who had undergone magnetic resonance imaging (MRI) under general anesthesia as a preparation for Fontan completion. Of the 21 children, 10 had HLHS and 11 had other forms of single ventricle physiology without systemic outflow obstruction or aortic coarctation (tricuspid atresia with ventriculoarterial concordance in 4, pulmonary atresia in 4, double-outlet right ventricle in 2, and double inlet left ventricle in 1). All patients had received a bidirectional superior cavopulmonary anastomosis as a part of their staged surgical management at the time of the study. All patients with HLHS had undergone standard surgical reconstruction of the aortic arch with a homograft patch at the Norwood procedure, with all procedures performed by only 2 surgeons. None of the other 11 children with other forms of single ventricle physiology had undergone any ascending aorta or aortic arch surgery, and they represented the control group.

MRI Data

All patients underwent MRI (1.5-T Avanto; Siemens Medical Solutions, Erlangen, Germany) before total cavopulmonary connection Fontan completion. Phase-contrast MRI was used for flow quantification. Through-plane flow data (about 30 ms temporal resolution, interpolated to 30 frames per heart beat) were acquired with the use of retrospective cardiac gating. Also, retrospectively gated, balanced, steady-state free precession cine images of the heart were acquired in the vertical long-axis, 4-chamber, and short-axis view covering the entirety of both ventricles (9-12 slices). Post-processing was performed using OsiriX (Pixmeo, Geneva, Switzerland) with in-house–written plug-ins. Single ventricle end-diastolic and end-systolic volumes were measured. The stroke volume (SV), ejection fraction, and cardiac output were calculated from these measurements. The ventricular volumes and cardiac output were indexed for body surface area. During the cardiac MRI scan, the blood pressure was measured every 5 to 10 minutes noninvasively with a cuff of appropriate size placed around the right arm or right leg. The systolic and diastolic blood pressure at the beginning of the scan was recorded, and the pulse pressure was calculated as the systolic minus the diastolic blood pressure.

Care was taken to ensure that, in the HLHS group, the imaging plane was always above the aortic–pulmonary anastomosis to ensure that the reconstructed aorta was captured. In all cases, the ascending aortic data was acquired just above the Damus-Kaye-Stansel anastomosis using a standard method, according to our institutional pre-Fontan completion protocol.

Institutional ethical approval for the retrospective use of the MRI data was obtained, and all patients' parents gave informed consent for research use of the imaging data.

Calculation of Aortic Distensibility

All calculations were semiautomated by the use of an in-house–written OsiriX plug-in.[13]

Aortic distensibility (D) was derived directly from knowledge of wave speed (c), according to the Bramwell-Hill equation:

$$D = \frac{1}{\rho c^2} \tag{1}$$

where ρ is the blood density. The calculation of wave speed (c) was determined from the area (A) and velocity (U) information, directly derived from the MRI data. The ascending aortic MRI flow data were segmented using a validated semiautomatic registration-based segmentation algorithm,[14] allowing extraction of the U and A signals. Changes in velocity (dU) and fractional changes in area (dlnA) were related using the water hammer equation:

$$dU_{\pm} = \pm c d\ln A_{\pm} \tag{2}$$

According to this equation, the U-lnA relationship should have a linear slope yielding the wave speed in early systole, when no backward waves are expected. This holds true for noncoronary arteries. The aortic U-lnA relationship is represented by a loop, similar to the pressure-velocity loop[15] and diameter-velocity loop[16] methods for wave speed calculation. All analyses were performed by the same operator.

Distensibility was also assessed in terms of stress (σ) and strain (ε) to evaluate the effect of the presence of the patch. Variations in diameter, d (Δd) over the diastolic diameter (d_{dias}) yielded an indication of circumferential strain ($\varepsilon = \Delta d/d_{dias}$), and circumferential stress was defined as follows:

$$\sigma = \frac{Pd}{2h} \tag{3}$$

where P is the pulse pressure, and h is the wall thickness.

Wave Intensity Analysis

Although the formulation of wave intensity is traditionally determined from the pressure and velocity[17] and, more recently, from diameter and velocity,[18] we have proposed a formulation using area and velocity.[13] Using the area and velocity, the net wave intensity (dI_A) and separated wave intensity ($dI_{A\pm}$) are respectively defined as follows:

$$dI_A = dU d\ln A \tag{4}$$

and

$$dI_{A\pm} = \pm\frac{c}{4}\left(d\ln A \pm \frac{1}{c}dU\right)^2 \tag{5}$$

Having obtained a dI_A pattern, the waves can be identified as compression ($d\ln A > 0$) or expansion ($d\ln A < 0$) waves. Depending on whether they are traveling away or toward the heart, the waves can also be defined as forward-traveling or backward-traveling, respectively. Traditionally, left ventricular (LV) ejection is described by a forward compression wave (FCW), and LV relaxation is described by a forward expansion wave (FEW). It has been shown in normal subjects that FCW correlates significantly with a maximum rate of pressure rise (dP/dt) and that the FEW correlates significantly with the LV relaxation time constant.[19] On this basis, the peak intensity of, and the energy carried by, the FCW and FEW were calculated. Wave energy (I) was derived from the area under the wave intensity curve, as follows:

$$I = \int_{t1}^{t2} dI_A dt \tag{6}$$

where t_1 and t_2 indicate, respectively, the onset and arrival of each wave. The energy carried by the FCW and FEW was also indexed for the SV, as an indication of the energy carried by a single unit of ejected volume.

It should be noted that the units of wave intensity and wave energy are not the traditional [W/m^2] and [J/m^2] but rather [m/s] and [m], respectively.

Albeit not intuitive, these units result from formulating wave intensity analysis in terms of area. The significance of the parameters is not altered.[18]

Statistical Analysis

The data are presented as the mean ± standard deviation. The differences in demographic and MRI variables between the HLHS and control groups were assessed with unpaired *t* test. Linear regression analysis was also performed to establish the correlations between the parameters.

RESULTS

The baseline characteristics of the HLHS and control group are listed in Table 1. No difference was seen in gender or body surface area between the 2 groups; however, patients with HLHS exhibited a lower SV ($P = .04$) and lower ejection fraction ($P = .003$). No patient had any evidence of aortic recoarctation. The arterial pulse pressure was similar in both groups ($P = .12$).

A sample of the U-lnA loop is shown in Figure 1, *A*, highlighting the slope yielding the wave speed value. The estimate of local ascending aortic wave speed showed greater values in the HLHS cohort (7.2 ± 2.4 vs 4.3 ± 0.9 m/s, $P = .002$). This resulted in a twofold less distensible aortic arch in the HLHS group, with D = 3.5 ± 2.9 × 10^{-3} 1/mm Hg compared with D = 7.8 ± 3.7 × 10^{-3} 1/mm Hg in the control group ($P = .004$; Figure 2).

Dimensional data derived from the minimal diastolic area at the level of the distal ascending aorta, assuming a circular shape, revealed that those patients with reconstructed aortic arches had a substantially larger aortic diameter than controls (2.1 ± 0.2 cm for the HLHS group vs 1.6 ± 0.2 cm for the control group, $P < .0001$). As expected, the aortic diameter and distensibility were significantly and inversely correlated ($R = .66$, $P = .001$), but no association was noted between the distensibility and indexed SV ($R = .02$, $P = .93$). Patients with HLHS, with a larger ascending aortic diameter, also exhibited 50% reduced ε compared with the control patients (0.10 ± 0.02 vs 0.19 ± 0.05, $P < .001$).

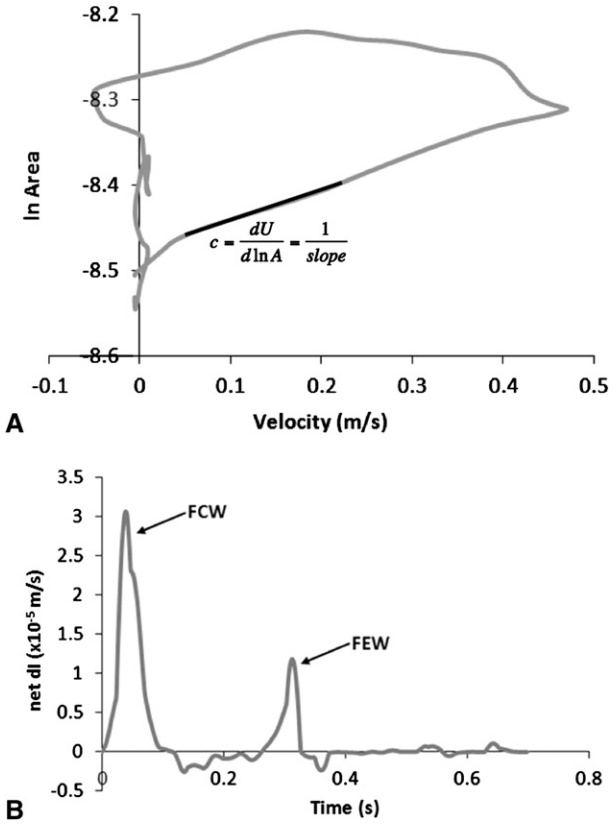

FIGURE 1. Outputs of the plug-in used for magnetic resonance imaging analysis. A, Sample of velocity *(U)* and ln area *(lnA)* loop for calculation of wave speed *(c)*. The inverse of the slope of the linear part of the loop in early systole yields c. B, Sample of wave intensity pattern from 1 patient showing the forward compression wave *(FCW)* and forward expansion wave *(FEW)* associated with ventricular ejection and relaxation, respectively.

TABLE 1. Patient characteristics

Variable	HLHS	Control	*P* value
Patients (n)	10	11	—
Gender (n)			.98
Male	6	7	
Female	4	4	
Age (y)	3.4 ± 1.0	4.6 ± 1.5	.02
Weight (kg)	14.5 ± 2.5	15.9 ± 4.2	.19
Height (cm)	95.7 ± 5.5	99.5 ± 9.7	.15
Body surface area (m^2)	0.61 ± 0.06	0.65 ± 0.11	.15
Heart rate (bpm)	107 ± 17	102 ± 12	.21
Ventricular ejection fraction (%)	52 ± 6	61 ± 6	.003
SV (mL)	29.9 ± 5.0	40.9 ± 18.1	.04
Ventricle type (n)			.0002
Right	10	2	
Left	0	9	
Aortic pressure pulse (mm Hg)	39	50	.12

HLHS, Hypoplastic left heart syndrome; *SV*, stroke volume.

Wave intensity exhibited a typical pattern, with FCW in early systole, reflected as a backward compression wave, and a FEW in late systole (Figure 1, *B*). Patients with HLHS showed a lower average FCW peak dI_A than the controls (8.8 ± 5.9 × 10^{-6} vs 34.5 ± 16.8 × 10^{-6} m/s, $P = .0002$) and lower energy carried by the FCW (3.7 ± 2.3 × 10^{-7} vs 12.0 ± 4.6 × 10^{-7} m, $P < .0001$). The same was observed for the FEW. The patients with HLHS showed a lower average FEW peak dI_A than controls (4.5 ± 2.9 × 10^{-6} vs 11.2 ± 5.1 × 10^{-6} m/s, $P = .001$) and lower wave energy (2.3 ± 1.4 × 10^{-7} vs 4.9 ± 1.5 × 10^{-7} m, $P = .0004$). These results are summarized in Figure 3.

The patients with HLHS appeared to be less energy efficient, because the FCW and FEW both carried significantly less energy per SV unit compared with control patients (13 ± 9 × 10^{-9} vs 33 ± 15 × 10^{-9} m/mL, $P = .001$; and 8 ± 5 × 10^{-9} vs 14 ± 7 × 10^{-9} m/mL, $P = .03$,

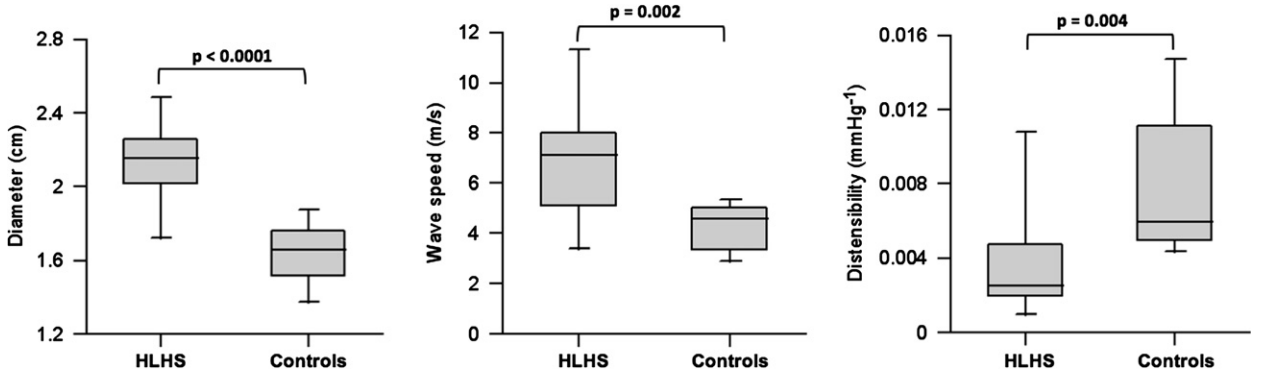

FIGURE 2. *Left*, Diameter, *center*, wave speed, and *right*, distensibility differences between hypoplastic left heart syndrome *(HLHS)* and control patients.

respectively). As shown in Figure 4, *A*, reduced FCW and FEW wave energy was not significantly related to the SV ($R = .41$, $P = .07$; and $R = .22$, $P = .37$, respectively). However, a much more significant correlation was identified between wave energy and diameter, for both FCW ($R = .82$, $P < .0001$) and FEW ($R = .74$, $P = .001$; Figure 4, *B*).

DISCUSSION

The present noninvasive study used cardiovascular MRI alone to assess the central aortic properties and, using MRI-derived wave intensity analysis, ventricular–vascular coupling. The findings indicated that aortic arch reconstruction using a patch at the Norwood procedure results per se in reduced distensibility in the ascending aorta and reduced efficiency of the ventricular–vascular coupling, which adds to the burden of increased afterload that the single ventricle must sustain long term. Changes in wave intensity peaks and the energy carried by the FCW and FEW in early and late systole, respectively, were taken as surrogate parameters suggesting abnormal coupling.

Previous studies using echocardiography or MRI have shown that patients with HLHS have reduced distensibility

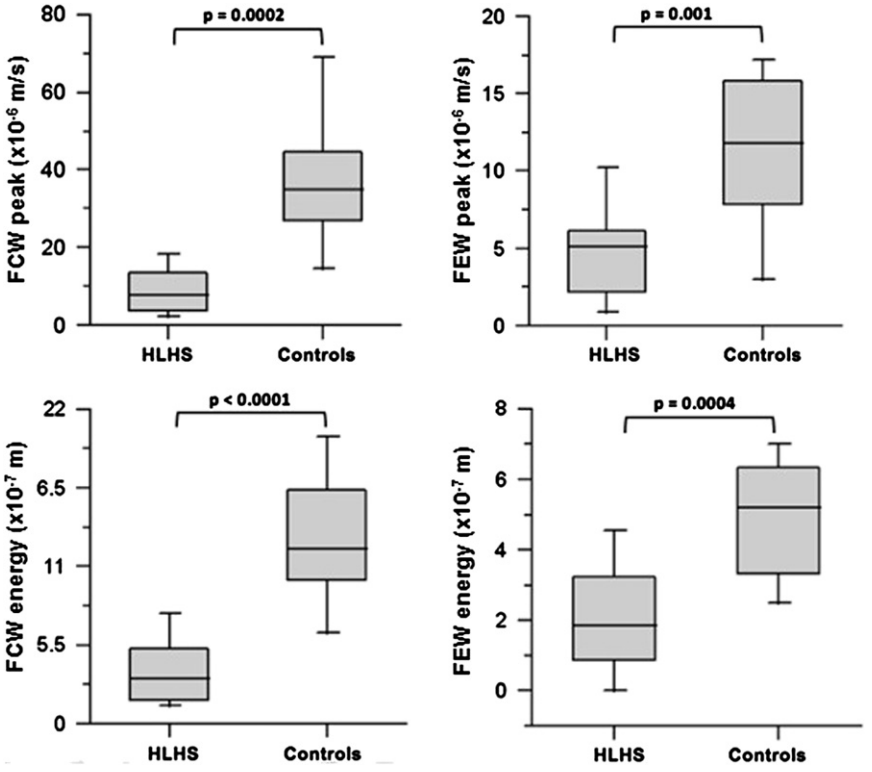

FIGURE 3. Patients with hypoplastic left heart syndrome *(HLHS)* exhibited marked differences with respect to control patients in terms of wave intensity results. Both the peaks of, and the energy carried by, the forward compression wave *(FCW)* and forward expansion wave *(FEW)* were significantly lower in the HLHS group.

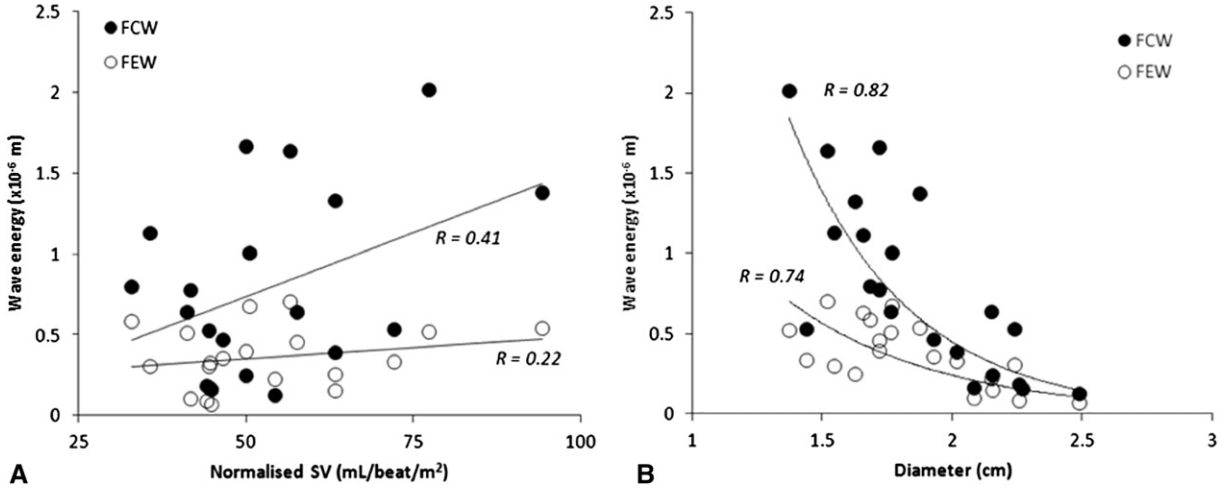

FIGURE 4. Correlation between (A) wave energy and indexed stroke volume and (B) wave energy and ascending aorta diameter, for both the forward compression wave *(FCW)* and forward expansion wave *(FEW)*. *SV,* Stroke volume.

of the ascending aorta compared with normal subjects.[11] Our data have confirmed that patients with HLHS have reduced distensibility of the proximal aorta; however, compared with the previous MRI report,[11,12] our control group included patients with single ventricle physiology in whom aortic reconstruction had not been performed. Importantly, the noninvasive MRI method we describe also gives insight into the mechanistic explanation for the reduction in distensibility (aortic dimensions vs material properties of the aorta) and energetic efficiency of the 2 different systems (single ventricle coupled to reconstructed aorta vs single ventricle with native aorta).

With regard to the aortic characteristics, our results have demonstrated an association between the diameter of the ascending aorta and distensibility. Such an association is not surprising and a relationship between worsening elastic properties and increasing size has been demonstrated in different aortic diseases.[20] However, our data suggest that the increased stiffness of the ascending aorta in HLHS is related not only to dilation of the ascending aorta but also to the intrinsic mechanical properties of the wall (ie the presence of a patch). By assessing the elastic properties of the aortic wall, we have shown that aortic wall strain was reduced by 50% in patients with HLHS compared with that in control patients. If the presence of the patch was mechanically negligible, patients with HLHS and controls would have to behave similarly. Given the difference in ε, this would only occur if the aortic pulse pressure in the HLHS group was 3 times larger than that in the control group. However, the observed pulse pressures were similar (Table 1); therefore, the influence of the patch is not negligible in terms of the effect on the mechanical properties of the aorta.

Clinically, the elasticity of the ascending aorta is important in determining the afterload on the systemic ventricle. Reduced distensibility results in greater aortic impedance, and with time, it can lead to changes in ventricular-aortic coupling.[21] This is known to negatively affect the ventricular diastolic function and can result in ventricular hypertrophy first[11] and ventricular systolic failure later. Our data support these observations. Patients with HLHS exhibited both reduced ejection fraction and SV. Moreover, as patients with HLHS age, the increased arterial stiffness and hypertension associated with normal aging might exacerbate the maladaptation of the systemic right ventricle. Such changes are also of particular concern because of the effect they might have on the transplanted heart.

Patients with HLHS, even after a perfect arch reconstruction without any residual aortic arch narrowing, exhibited lower intensity of both FCW and FEW, indicating compromised ventricular systolic and diastolic function, with the FCW being associated with dP/dt and the FEW associated with the ventricular relaxation coefficient τ.[19] Hence, in our study, the energy carried forward by each beat was lower in those with HLHS, even after correcting for differences in SV, and this might reflect unfavorable ventricular–vascular coupling compared with other patients with single ventricle physiology who have not undergone aortic arch surgery.

In the present setting, loss of wave energy could potentially be a consequence of abnormalities on the ventricular side of the ventricular–vascular unit (ie, ejected SV) and/or abnormalities on the arterial side (ie, aortic diameter, presence of the patch), causing increased afterload. We have shown that distensibility and wave energy are not significantly dependent on SV. However, a significant association between the aortic diameter and both distensibility and wave energy was identified. Although a larger than normal dimension of the reconstructed ascending aorta and aortic arch is a common feature of the Norwood operation, the increased size of the arch and the consequent ‘‘pooling’’ of

blood in the ascending aorta appear to be detrimental to the mechanical performance of the single ventricle and the ventriculoarterial coupling. This is particularly concerning given the already increased aortic impedance observed in other single ventricle patients after completion of the Fontan operation.[9]

From a methodologic perspective, the proposed implementation of wave intensity analysis derived from a single MRI measurement has the advantage of being pressure independent, noninvasive, and performed at the specific location of interest, in this case in the ascending aorta. The presented wave intensity results and their implications for compromised ventriculoarterial coupling in patients with HLHS are overall in agreement with previous pressure-based investigations.[11,12]

Previous studies focusing on distensibility quantification necessitated invasive arterial pressure monitoring, combined with area change information.[12] In the absence of invasive catheter measurements, cuff pressure data have been used for distensibility calculations.[14] Alternatively, distensibility has been derived from pulse wave velocity estimated with MRI examination.[22] This MRI-based method requires multiple acquisitions and foot-to-foot (transit time) wave speed calculation. We propose a completely noninvasive and semiautomated method based solely on analysis of a single MRI slice for estimation of local distensibility.

Clinical Perspective

Our results suggest that limiting the diameter of the reconstructed proximal aorta at the Norwood operation and the development and use of materials with better long-term elastic properties to reconstruct the ascending aorta and the aortic arch might prove beneficial for patients with HLHS, although initial surgery is also about ensuring a large-enough aorta to account for patient growth. Data on aortic distensibility and ventricular–vascular coupling from other institutions using materials other than the aortic homograft for aortic reconstruction would be useful to detect any difference in elastic properties with the use of different materials. We acknowledge that although the ascending aorta appears to be larger than normal in Norwood patients after stage II, no longitudinal information is available from our study cohort regarding progressive growth, and it is possible that surgical enlargement of the ascending aorta at the Norwood stage I procedure is an unavoidable requirement to achieve a normal adult aortic size.

Wave intensity analysis provides a measure of the power transported by the waves propagating within the arteries,[23] and, as such, it is suitable to assess the working condition of the heart interacting with the arterial network.[24] It can be shown that, theoretically, peak FCW and peak FEW are related to maximal rate of LV pressure increase and maximum aortic deceleration toward end ejection, respectively.[25] This is also supported by in vivo observations, linking the FCW and FEW to parameters of ventricular function.[19,25] Thus, this method is well suited to gather mechanical and energy considerations in the selected patients.

Although we report the use of noninvasive techniques to investigate ventriculoarterial interaction, we acknowledge that measurement of the pressure–volume loops from conductance catheters remains the reference standard for studying ventriculoarterial coupling. Preliminary data[26] have shown that arterial wave intensity correlates with the ventricular/vascular elastance ratio, supporting the clinical relevance of the method. In this case, dI was calculated using the dPdU formulation. This comparison warrants additional study.

Study Limitations

In the present study, arterial blood pressure was measured noninvasively with cuffs placed on either the arm or leg, depending on the position of the venous line used for gadolinium infusion. We believe that even if not ideal, this was not a significant limitation of our study for at least 2 reasons. First, the presence of recoarctation as a cause of a lower blood pressure in the leg in some patients could be excluded in those subjects who had undergone previous aortic surgery. Second, the comparison between the pulse pressure measured in children with versus without a reconstructed arch was only used to suggest that the latter would require a pulse pressure 3 times greater than that observed in HLHS for their aorta to behave along the same stress–strain relationship.

As per other noninvasive formulations of wave intensity,[16] the dI units in the present study were not intuitive compared with the traditional [W/m^2] and, thus, slightly obscured the physical significance of the calculated parameters. However, the underlying significance of the waves, discussed in other studies,[24] remains.

CONCLUSIONS

A novel method involving MRI-derived wave intensity analysis was used to link the effect of arch reconstruction with compromised ventriculoarterial coupling. The present results indicate that arch reconstruction at the Norwood procedure results in reduced ascending aorta distensibility, adding to the increased afterload that the systemic single ventricle must face chronically. Our data suggest that the use of smaller patches and/or patches made from different materials warrants additional investigation, because they could be potentially more beneficial from a mechanical and energy perspective.

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Discussion

Dr William M. DeCampli *(Orlando, Fla).* Thank you. This was a very well-presented paper, Dr Giardini.

I applaud the authors' use of high-resolution MRI data for application to wave intensity analysis. They have nicely corroborated evidence previously published in 2010 showing that the Norwood arch becomes stiffer than the normal aorta. This method is actually a very accurate way of determining the vascular wall stiffness.

The authors used these data, however, to make some rather broad inferences about ventricular function. The reference standard for quantifying intrinsic myocardial function, ventricular arterial coupling, and myocardial energy efficiency is the determination of ventricular end-systolic elastance and effective arterial elastance from ventricular pressure–volume loops. These quantities cannot be determined with wave intensity alone. Given that, combining the authors' technique with the acquisition of instantaneous ventricular pressure data could yield powerful information about the interpretation of the waves seen in wave intensity analysis and a new level of understanding of single ventricle function.

I have 3 questions, as follows:

The physical meaning of your ''wave intensity'' is somewhat abstruse, because it is not really an energy flux but seems to have units of meters cubed per second. This quantity is not very intuitive. You can calculate the more traditional energy flux from your data, however, because there is a direct, in fact, causal relationship between the changes in pressure and cross-sectional area. How does this more physically intuitive quantity compare between your 2 groups?

Dr Giardini. We do not have invasive pressure data to be able to calculate the standard wave intensity, 1 of the limitations of our study. However, we believe it is probably also 1 of the strengths, because we were actually able to collect this information completely noninvasively. In particular, we were able to collect it at the specific point of interest in the ascending aorta, which I believe is another strength. With this technique, we are able to measure the wave intensity at different levels in the aortic arch, for example, if we are interested. As you pointed out, traditionally, calculation of wave intensity analysis is determined from the pressure and flow velocity data, and the result is a value expressed as watts divided by square meters. We have proposed a new formulation of wave intensity analysis using the area and flow velocity, which has a very sound basis in the conservation of mass and conservation of momentum. Also, because we are actually using the area and

velocity as variables, the results will be expressed in this unit. Thus, they do not necessarily translate in one of each other, but they are a representation of the same phenomenon.

Dr DeCampli. In 2005, Nakayama and coworkers empirically showed that wave intensity was, in fact, preload dependent. Specifically, they showed that if one indexed the wave intensity to the square of the end-diastolic volume, one achieved the best-fit empirical correlation with ventricular end-systolic elastance.

Now, you indexed your wave intensity to the SV. What physical rationale did you have to do this, and how do you know that was the appropriate method to index it?

Dr Giardini. We are aware of the report you cited and actually because of that, we have been considering whether we should index wave intensities according to ventricular volumes. We actually have that information. However, a large body of evidence suggests that wave intensity analysis results that are not adjusted for any preload dependence seem to be very closely related to invasively measured dP/dT and with the LV relaxation coefficient τ; thus, we were not compelled by that need to index our data for preload. The only adjustment we made was indexing the wave energy for SV because we had some degree of correlation with SV. We wanted to create a variable that could be, to a certain degree, independent of the SV and would express how efficiently the ventricle is able to eject blood and which energy is associated with each milliliter of ejection produced.

Dr DeCampli. Finally, in wave analysis of the Norwood group, one would expect to see reflected or backward waves, in addition to forward waves, and very probably by the end of systole. These are the waves that actually add to ventricular afterload and can adversely affect ventricular function. Where were the reflected waves in your analysis?

Dr Giardini. We have not analyzed those data, but actually in 1 of the graphs that I showed some reflected waves were represented, which actually tended to occur quite early in systole. The study of wave reflection with this technique is our next area of interest. We will return and study the reflected waves, especially at this time, including patients with aortic coarctation in whom we expect much wave reflection.

Dr J. William Gaynor *(Philadelphia, Pa).* Thank you, I enjoyed that very much.

Which part of the arch is responsible for most of the effect? In particular, how much did geometry affect it? Because we frequently see that it is bulbous in the transverse arch, narrows down, and is smaller in the ascending. Is there a specific part of the arch, or is it a sum of everything? How we tailor the arch might be as important as what material we use. Unfortunately, it always seems bulbous in some places along the arch.

Dr Giardini. You are correct that for practical purposes, if we are thinking about how to make that part of the circulation better, it is very important to consider the size and the shape of the aortic arch. Most of the effect we have seen seems to be due to the size of the proximal ascending aorta and less to the nature and characteristic of the patch, although we could demonstrate that the patch has an effect, because the 2 arteries are just behaving differently beyond the effect of the difference in size.

Dr Christopher A. Caldarone *(Toronto, Ontario, Canada).* I would like to ask a follow-up question to that. By measuring the distensibility in the ascending aorta, it is necessarily adding the component of the patch and the native aorta, together, in the individual patient and you are comparing that with controls who do not have a patch. However, you are not able to address the question of whether the aorta itself is less distensible or more distensible. But you could do so if you compared the proximal descending thoracic aorta in both groups. Have you thought of doing that analysis?

Dr Giardini. We have data regarding that, and we were actually able to demonstrate that the descending thoracic aorta has normal elastic properties in HLHS; therefore, probably it is not a diffuse arterial disease as far as we understand it.

Dr Shunji Sano *(Okayama, Japan).* I reconstructed the neoaorta, and most of the time in direct anastomosis without any homograft or foreign materials. From your conclusion, is this technique different from the homograft patch to reconstruct the neoaorta?

Also, I am always just worried about the size. When I reconstruct the neoaorta without any patch, the ascending aorta is quite large, and then, suddenly, the descending aortic size is down. If you have a large homograft, the size gradually decrease, changes. Is there any difference in this discrepancy, distensibility, or is there an energy loss of the size?

Dr Giardini. We do not have that data in our population, because all our patients were treated uniformly with the use of a homograft patch. Thus, we do not have any patient who has undergone that type of Norwood operation without any patch material. I think there could be an advantage in doing that. Something else we were not able to estimate was the presence of a circumferential suture line, which clearly has to have also an effect, especially in terms of wave reflections.